Peak flowmeter resistance decreases peak expiratory flow in subjects with COPD

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Miller, Martin R., and Ole F. Pedersen. Peak flowmeter resistance decreases peak expiratory flow in subjects with COPD. J Appl Physiol 89: 283–290, 2000.—Previous studies have shown that the added resistance of a mini-Wright peak expiratory flow (PEF) meter reduced PEF by ~8% in normal subjects because of gas compression reducing thoracic gas volume at PEF and thus driving elastic recoil pressure. We undertook a body plethysmographic study in 15 patients with chronic obstructive pulmonary disease (COPD), age 65.9 ± 6.3 yr (mean ± SD, range 53–75 yr), to examine whether their recorded PEF was also limited by the added resistance of a PEF meter. The PEF meter increased alveolar pressure at PEF (Ppeak) from 3.7 ± 1.4 to 4.7 ± 1.5 kPa (P = 0.01), and PEF was reduced from 3.6 ± 1.3 l/s to 3.2 ± 0.9 l/s (P = 0.01). The influence of flow limitation on PEF and Ppeak was evaluated by a simple four-parameter model based on the wave-speed concept. We conclude that added external resistance in patients with COPD reduced PEF by the same mechanisms as in healthy subjects. Furthermore, the much lower Ppeak in COPD patients is a consequence of more severe flow limitation than in healthy subjects and not of deficient muscle strength.

Key words: peak flow determining factors; thoracic gas compression; added expiratory resistance; chronic obstructive pulmonary disease; peak expiratory flow

It has recently been shown that the added resistance of a mini-Wright peak flowmeter decreases the achieved peak expiratory flow (PEF) by ~8% compared with the PEF measured by a pneumotachograph (PT) (10). Furthermore, there is evidence that PEF is determined by the wave-speed flow (Vws)-limiting mechanism in most healthy subjects and possibly in asthmatic subjects (9). Fry and Hyatt (5) observed that an added resistance shifts the isovolume pressure-flow curves to the right. The alveolar pressure at PEF (Ppeak) must therefore be increased by the added pressure drop across the PEF meter. The total pressure necessary to reach a given PEF may therefore be so large that it cannot be achieved even with maximum effort. This possibility has already been explored in 10 healthy men studied in a body plethysmograph with facilities to measure Ppeak as well as mouth flow vs. expired volume and displaced thoracic gas volume (TGV) (10). It was found that PEF in healthy subjects obtained with an added resistance such as a mini-Wright meter was wave-speed determined, but the added resistance caused gas compression, and the resulting lower TGV at PEF was the reason that the PEF was lower. In patients with severe airway obstruction and emphysema, however, the situation may be different. They have a low Ppeak (1) and a large TGV. The lower Ppeak will lead to less gas compression in the lungs of the emphysematous patients, but this may be counterbalanced by the larger TGV, which would allow a greater absolute change in volume due to compression. Furthermore, due to destruction of lung tissue, the elastic recoil pressure of the lungs (Pel) is smaller for a given lung volume than in healthy subjects, and the relationship between TGV and Pel is different. Because Pel is a main determinant of maximum flow (Vmax) (8), we cannot immediately predict how gas compression in the lungs will affect PEF in these patients. The aim of this study, therefore, is to examine how added resistance influences PEF in patients with obstructive lung disease and how the resistance of a mini-Wright peak flowmeter influences PEF.

MATERIALS AND METHODS

Equipment. We used a slightly modified pressure-corrected flow body plethysmograph with a 650-liter constant volume box. Box pressure was measured across a 64-cm² single-layer 400-mesh screen in the wall of the plethysmograph by use of a Validyne MP45 transducer with a ±0.2-kPa membrane to provide flow in and out of the box. Mouth flow was measured by use of an unheated Fleisch-type PT with a diameter of 5 cm, normally used by Vitalograph Compact spirometer (Buckingham, UK). The pressure across the PT head was measured by a similar Validyne transducer. The PT head was provided with a conical inlet containing a wire screen and was tested as described previously (12) with steady and dynamic flows. It was linear up to 16 l/s; R² = 0.9994, residual standard error = 2.1 l/s.
deviation (RSD) from regression line through the origin = 0.11 l/s. The signals from the transducers were low-pass filtered at 40 Hz (18 dB per octave), fed to a computer via an analog-to-digital conversion board (Dash-16, Metrabyte, Taunton, MA) with the data acquisition and calculation being performed by Asyst software (version 1.56, McMillan). The sampling rate was 200 Hz with a total sampling time of 5 s.

Calibration was performed by using an explosive decompression device able to deliver 7.79 liters (11), with a PEF of 2.36 liters per second and a 10–90% PEF rise time of 40 ms. Gas flow from the decompression device entered the box via the mouth PT, which during calibration was inverted so that the flow was in the expiratory direction. The delivered flow continued through the box and out of the screen. Mouth flow was calibrated from volume and time. Damping of the box flow due to the capacitance of the air in the box was corrected for by adding a signal proportional to the first derivative of box pressure vs. time to the pressure signal used in the box flow recording (15). Box flow was displayed against mouth flow, and the correction was adjusted so that a closed loop was obtained. Box flow was matched to mouth flow by using a third-degree polynomial, making the reading of the box flow identical to that of the mouth flow during the calibration procedure. Electronic and thermal drifts were carefully corrected before each measurement.

Determination of alveolar pressure. Because of the compression of alveolar gas during a forced expiration, the amount of air leaving the box from the mouth via the PT will be less than the amount entering the box via the screen. This difference between the expired mouth volume and the volume entering the box (ΔV) changes throughout expiration, being zero at the start of the blow and at the finish when the subject relaxes. With isothermal conditions assumed, alveolar pressure (PA) can be determined by application of Boyle’s law

$$P_A = \frac{\Delta V(P_B - P_{H2O})}{(TLC - Vb)}$$  \hspace{1cm} (1)

where $P_B$ is the barometric pressure, $P_{H2O}$ is the pressure of saturated water vapor at 37°C, TLC is the total lung capacity from which the expiration starts, and Vb is the volume entering the box during the expiration. This equation is essentially the same as that applied by Ingram and Schilder (7) and Zamel et al. (16).

Validation of pressure measurements. The validity of such pressure calculations was examined in our previous study (10). A servo-controlled calibration pump recently developed at the University of Birmingham was placed in the plethysmograph. An artificial flow profile was delivered by the pump with 9.99 l/s PEF, 30 ms time from 10 to 90% PEF, and 20 ms duration of flow above 90% PEF. The pump expired through stiff tubing (3.5 cm internal diameter) that was connected to a slide valve with orifices of different sizes, which in turn was connected to the mouth PT. With no extra resistance, the pressure drop across the mouthpiece assembly depended on flow (V) as described by the equation pressure = 0.0148(V)^1.57 kPa ($R^2$ = 0.9996, RSD = 2.1%). In place of the slide valve, a mini-Wright PEF meter was also used as an added resistance. The meter was enclosed in a Perspex holder so that the air passing through the variable-orifice

![Diagram](image-url)  
Fig. 1. Mouth flow vs. expired volume for forced expiratory maneuvers through added resistances (Orif 0: no added resistance, Orif 2: 10.5-mm orifice, Orif 3: 8.5-mm orifice, Orif 4: 6.5-mm orifice). PEF meter, peak expiratory flowmeter; %FVC, percent forced vital capacity. Means are shown for 15 subjects (see text). See Table 2 for details.

Table 1. Baseline lung function data for subjects with chronic obstructive pulmonary disease

<table>
<thead>
<tr>
<th>Subject No.</th>
<th>Gender</th>
<th>PEF, l/s</th>
<th>PEF, %pred</th>
<th>FEV₁, liters</th>
<th>FEV₁, %pred</th>
<th>FVC, liters</th>
<th>FVC, %pred</th>
<th>TLC, liters</th>
<th>TLC, %pred</th>
<th>$TlCO$, mmol · min⁻¹ · kPa⁻¹</th>
<th>$TlCO$, %pred</th>
</tr>
</thead>
<tbody>
<tr>
<td>1</td>
<td>M</td>
<td>3.42</td>
<td>44</td>
<td>0.98</td>
<td>35</td>
<td>3.05</td>
<td>82</td>
<td>6.93</td>
<td>105</td>
<td>4.39</td>
<td>52</td>
</tr>
<tr>
<td>2</td>
<td>F</td>
<td>1.58</td>
<td>32</td>
<td>0.68</td>
<td>45</td>
<td>1.65</td>
<td>89</td>
<td>5.73</td>
<td>125</td>
<td>4.16</td>
<td>70</td>
</tr>
<tr>
<td>3</td>
<td>M</td>
<td>2.00</td>
<td>27</td>
<td>0.53</td>
<td>20</td>
<td>1.72</td>
<td>51</td>
<td>8.80</td>
<td>154</td>
<td>4.52</td>
<td>57</td>
</tr>
<tr>
<td>4</td>
<td>F</td>
<td>3.08</td>
<td>60</td>
<td>0.84</td>
<td>52</td>
<td>1.73</td>
<td>88</td>
<td>4.51</td>
<td>108</td>
<td>5.99</td>
<td>97</td>
</tr>
<tr>
<td>5</td>
<td>M</td>
<td>2.42</td>
<td>31</td>
<td>0.69</td>
<td>24</td>
<td>1.60</td>
<td>41</td>
<td>6.85</td>
<td>100</td>
<td>4.99</td>
<td>57</td>
</tr>
<tr>
<td>6</td>
<td>M</td>
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<td>35</td>
<td>0.93</td>
<td>37</td>
<td>3.02</td>
<td>93</td>
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<td>111</td>
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<td>53</td>
</tr>
<tr>
<td>7</td>
<td>F</td>
<td>1.50</td>
<td>25</td>
<td>0.43</td>
<td>19</td>
<td>1.77</td>
<td>66</td>
<td>6.17</td>
<td>135</td>
<td>3.00</td>
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<tr>
<td>8</td>
<td>M</td>
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<td>1.03</td>
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<td>6.76</td>
<td>105</td>
<td>7.76</td>
<td>91</td>
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<tr>
<td>9</td>
<td>F</td>
<td>1.92</td>
<td>36</td>
<td>0.54</td>
<td>29</td>
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<td>82</td>
<td>5.83</td>
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<td>85</td>
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<tr>
<td>10</td>
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<td>49</td>
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<td>46</td>
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<td>86</td>
<td>3.19</td>
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<td>7.52</td>
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<tr>
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<td>2.75</td>
<td>34</td>
<td>0.57</td>
<td>19</td>
<td>3.39</td>
<td>85</td>
<td>7.95</td>
<td>115</td>
<td>4.78</td>
<td>53</td>
</tr>
<tr>
<td>13</td>
<td>M</td>
<td>2.67</td>
<td>34</td>
<td>0.68</td>
<td>24</td>
<td>2.29</td>
<td>60</td>
<td>9.57</td>
<td>140</td>
<td>4.20</td>
<td>49</td>
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<td>14</td>
<td>M</td>
<td>4.58</td>
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<td>1.42</td>
<td>48</td>
<td>2.96</td>
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<td>66</td>
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<tr>
<td>15</td>
<td>M</td>
<td>2.58</td>
<td>33</td>
<td>0.97</td>
<td>34</td>
<td>3.53</td>
<td>91</td>
<td>8.93</td>
<td>128</td>
<td>4.21</td>
<td>48</td>
</tr>
</tbody>
</table>

Mean ± SD: 2.60 ± 0.79 38 ± 11 0.79 ± 0.26 33 ± 12 2.38 ± 0.74 74 ± 16 6.91 ± 15.6 119 ± 18 4.59 ± 1.36 59 ± 20

M, male; F, female; PEF, peak expiratory flow; %pred, percentage of predicted value; FEV₁, forced expiratory volume in 1 s; FVC, forced vital capacity; TLC, total lung capacity; $TlCO$, transfer factor for CO.
First, the TLC was determined by using a conventional plethysmographic technique (14). Then, while seated in the plethysmograph, the subjects performed at least three satisfactory FVC blows through orifices of different sizes and through the mini-Wright PEF meter. Flow-volume curves were recorded with mouth flow vs. mouth volume change and mouth flow vs. thoracic volume change. Furthermore, Pa was recorded vs. thoracic volume change. On-line calculation included PEF at the mouth (PEFm), PEF of thoracic volume change (PEF entering the box (PEFb)), mouth volume expired at PEFm, thoracic volume displaced at PEFm, thoracic volume displaced at PEFm, rise time to PEFm, rise time to PEFb, FVC at the mouth (FVCm), thoracic volume change (FVCb), and mouth flows at 5% intervals from 100% FVCm and 100% FVCb to low in the vital capacity. Furthermore, we determined by interpolation on the raw data the dwell times for mouth flow, that is, the duration of flow in excess of 90, 95, and 97.5% of PEFm. Pa was calculated for PEFm, and at 5% intervals of FVCb. Baseline spirometry was evaluated according to European standards (14). We used the same % increase as in the previous study (10). The influence of added resistance on PEF variability was calculated from the difference between the highest and next highest PEF values in each series of blows.

Statistics. The on-line calculated data were evaluated by use of the SPSS/PC statistical package (SPSS, Chicago, IL). Values are given as means ± SD. Because of the effect of gas compression on the flow-volume curve obtained at the mouth, we decided to use the mean values from the three satisfactory blows (replications) from each subject instead of applying any special selection criteria (14). Mean flow-volume curves and pressure-volume curves for all subjects were constructed for blows with the different added resistances by averaging the flows at identical percentages of FVC (%FVC). In these curves were incorporated mean peak flows plotted vs. mean %FVC. Paired differences were analyzed by a non-parametric test (Wilcoxon), and the overall differences between orifices were analyzed by ANOVA including the effects of orifice, replication, and subject. P = 0.05 was considered significant.

RESULTS

Test of calibration. PEF for the calibration mouth flow was 12.4 ± 0.2 l/s (n = 8); for the simultaneously

meter was collected and directed through the PT measuring mouth flow (12). This configuration increased the pressure across the mouthpiece assembly to 0.189(V)1.11 kPa (R² = 0.9996, RSD = 1.4%).

The volume reduction in the pump due to expiration and compression of air was replaced by air entering the box via the screen. Pa was directly measured in the pump by a Validyne transducer with a ±20-kPa diaphragm and was sampled in the same way as the mouth and box flow measurements after calibration with a mercury manometer at 10 kPa. Measured Pa at PEF was compared with Pa calculated from Eq. 1 and was multiplied by 1.4 to account for adiabatic conditions in the pump. This was done with and without the added resistances described above. It was found that the maximum error of derived pressure was an underestimate at the highest pressure (highest resistance) of 0.4 kPa. The percentage of absolute error was generally below 3%.

Subjects and measurements. Nineteen subjects with chronic obstructive pulmonary disease (COPD), who were attending the outpatient clinic of the University Hospital of Birmingham, gave informed consent according to the Helsinki declaration to participate in the study. Because of technical problems, curves from two of the patients were incorrectly recorded and could not be used. Two other subjects were not able to complete all the recordings. Data from the remaining 15 subjects who completed the study are presented. Baseline lung function data for these subjects, 5 women and 10 men, aged 65.9 ± 6.3 yr (mean ± SD, range 53–75 yr), were obtained before the study and are described in Table 1. PEF and forced expiratory volume in 1 s were all >2 standardized residuals (SR) below predicted values (14), i.e., below the lower 98% confidence limit, and so were clearly decreased for all subjects. This was true for forced vital capacity (FVC) only in six subjects. TLC was increased in all but one subject and by >2 SR in six (mean ± SD increase was 1.66 ± 1.48 SR). Transfer factor for carbon monoxide (diffusing capacity) was decreased in all subjects and by >2 SR in 11 subjects. It is clear from Table 1 that all the subjects had airflow limitation and most of them had changes consistent with emphysema.
Table 2. Effect of added resistance on PEF

<table>
<thead>
<tr>
<th></th>
<th>Control</th>
<th>PEF Meter</th>
<th>Difference from Control*</th>
<th>Orifice 2</th>
<th>Orifice 3</th>
<th>Orifice 4</th>
<th>ANOVA</th>
</tr>
</thead>
<tbody>
<tr>
<td>PEF&lt;sub&gt;m&lt;/sub&gt;, l/s</td>
<td>3.6 ± 1.3</td>
<td>3.2 ± 0.9</td>
<td>P = 0.01</td>
<td>2.9 ± 0.9</td>
<td>2.6 ± 0.7</td>
<td>2.0 ± 0.5</td>
<td>P &lt; 0.01</td>
</tr>
<tr>
<td>PEF&lt;sub&gt;b&lt;/sub&gt;, l/s</td>
<td>9.7 ± 3.0</td>
<td>10.0 ± 3.0</td>
<td>NS</td>
<td>9.6 ± 2.7</td>
<td>9.7 ± 2.5</td>
<td>9.1 ± 2.7</td>
<td>NS</td>
</tr>
<tr>
<td>Mouth volume expired</td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>at PEF&lt;sub&gt;m&lt;/sub&gt;, liters</td>
<td>0.14 ± 0.07</td>
<td>0.16 ± 0.12</td>
<td>NS</td>
<td>0.16 ± 0.07</td>
<td>0.15 ± 0.07</td>
<td>0.16 ± 0.09</td>
<td>NS</td>
</tr>
<tr>
<td>Thoracic volume</td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>displaced at PEF&lt;sub&gt;m&lt;/sub&gt;, liters</td>
<td>0.40 ± 0.13</td>
<td>0.48 ± 0.19</td>
<td>P &lt; 0.01</td>
<td>0.48 ± 0.17</td>
<td>0.52 ± 0.17</td>
<td>0.62 ± 0.23</td>
<td>P &lt; 0.01</td>
</tr>
<tr>
<td>Rise time to PEF&lt;sub&gt;m&lt;/sub&gt;, ms</td>
<td>75 ± 24</td>
<td>92 ± 59</td>
<td>NS</td>
<td>104 ± 52</td>
<td>98 ± 46</td>
<td>123 ± 39</td>
<td>P &lt; 0.01</td>
</tr>
<tr>
<td>Rise time to PEF&lt;sub&gt;b&lt;/sub&gt;, ms</td>
<td>76 ± 25</td>
<td>77 ± 41</td>
<td>NS</td>
<td>75 ± 28</td>
<td>77 ± 33</td>
<td>83 ± 28</td>
<td>NS</td>
</tr>
<tr>
<td>Dwell time at 90% PEF&lt;sub&gt;b&lt;/sub&gt;, ms</td>
<td>12 ± 4</td>
<td>12 ± 5</td>
<td>NS</td>
<td>17 ± 7</td>
<td>20 ± 7</td>
<td>35 ± 22</td>
<td>P &lt; 0.01</td>
</tr>
<tr>
<td>Alveolar pressure at PEF&lt;sub&gt;m&lt;/sub&gt;, kPa</td>
<td>3.7 ± 1.4</td>
<td>4.7 ± 1.5</td>
<td>P &lt; 0.01</td>
<td>4.6 ± 1.6</td>
<td>5.3 ± 1.8</td>
<td>6.6 ± 2.1</td>
<td>P &lt; 0.01</td>
</tr>
<tr>
<td>PEF variability, l/s</td>
<td>0.14 ± 0.17</td>
<td>0.18 ± 0.11</td>
<td>NS</td>
<td>0.12 ± 0.10</td>
<td>0.09 ± 0.11</td>
<td>0.07 ± 0.06</td>
<td>P &lt; 0.01</td>
</tr>
</tbody>
</table>

Values are means ± SD. PEF<sub>m</sub>, PEF measured at the mouth; PEF<sub>b</sub>, peak of flow entering the box (PEF of thoracic volume displacement); NS, not significant. Difference between control and PEF meter values was determined by Wilcoxon nonparametric test for pairs. Dwell time at 90% PEF<sub>m</sub> is duration of flow in excess of 90% PEF<sub>m</sub>. PEF variability is difference between two largest of three measurements (see text).

measured box flow, it was ~4% lower at 11.9 ± 0.2 l/s.

The rise time from 10 to 90% PEF<sub>m</sub> was 41 ± 2 ms, and that of the corrected box flow was 54 ± 4 ms. The fact that the loop could be closed and that the rise times recorded are close to the lower fifth percentile of rise times in a population study (11) indicate that the frequency response of the system was adequate.

**Flow-volume and pressure-volume curves.** Figure 1 shows mean mouth flows for all 15 subjects through the different added resistances, plotted against expired volume in %FVC<sub>m</sub>. The PEF points obtained with the added resistances occur within the perimeter of the flow-volume curves and are almost at the same expired volume.

Figure 2 similarly shows mean mouth flows plotted against TGV change in %FVC<sub>b</sub>. The PEF points now occur on the perimeter of the flow-volume curves or even exceed it, and with the higher resistances they occur at progressively lower TGV.

Figure 3 shows mouth flow vs. P<sub>A</sub>. Increased resistance causes the mouth PEF to decrease and the Ppeak to increase in a systematic manner. The curves are shown only for the increasing P<sub>A</sub> up to the achieved maximum pressure. If all the data are plotted, each curve loops back to the origin.

Because PEF of the different subjects did not occur at the same %FVC, the mean of the individual peak flows at the given %FVC exceeded the flow-volume and flow-pressure curve perimeters. Including the peak flow means therefore gives an unrealistic picture if points in the vicinity of PEF showing much lower flows were included. In Figs. 1–3, therefore, such neighboring points were not plotted.

In Table 2, the essential features from Figs. 1–3 and other calculations are summarized. Table 2 shows that PEF<sub>b</sub> is considerably larger than the PEF<sub>m</sub>. Also, PEF<sub>b</sub> and the rise time to PEF<sub>b</sub> are not influenced by increasing the external resistance. However, the rise time to PEF<sub>m</sub> tends to increase with resistance, indicating that a time lag between peak box flow and peak mouth flow progressively develops with increasing external resistance.

Addition of the mini-Wright PEF meter significantly reduced PEF<sub>m</sub>, whereas the expired mouth volume at PEF<sub>m</sub> did not change. However, the TGV displaced at PEF<sub>m</sub> significantly increased by 0.08 ± 0.10 liters (P = 0.01), so that PEF<sub>m</sub> occurred at a lower volume on the perimeter of the mouth flow vs. box-volume curve. The PEF meter caused the P<sub>A</sub> at PEF<sub>m</sub> to increase by 0.95 ± 0.77 kPa, which was not significantly different from the pressure drop across the PEF meter, determined by separate experiments to be 0.60 ± 0.17 kPa for the same values of PEF<sub>m</sub> (P = 0.09). The PEF meter did not increase the dwell time at 90% PEF<sub>m</sub>, but the dwell time increased when larger resistances were added, indicating blunting of the curves, which is not apparent in Fig. 1. The absolute PEF variability was not changed with the PEF meter, but higher resistances decreased the variability. However, when variability was expressed as a fraction of the mean PEF, there was no significant change in variability with increased resistance.

**DISCUSSION**

We have found that the PEF recorded with a mini-Wright PEF meter was reduced by an average of ~10% in patients with airflow limitation due to COPD compared with PEF recorded with a PT. This difference was due to the effect of gas compression in the lungs secondary to the imposed additional resistance and was not because the subjects’ ability to generate sufficient driving pressure was exceeded. These results have been derived from demanding plethysmographic
procedures, and the validity of these experiments must first be explored.

The initial calibration and tuning of the box was undertaken by sudden decompression delivering gas flow up to 12 l/s, which is larger than the mean PEFb from our subjects (Table 2). The rise time of the calibration signal was also comparable to the mean for our subjects, as shown in Table 2. As the mouth PT previously has been found to be linear up to at least 16 l/s, we believe that linearity, frequency response, and calibration procedure of the system were adequate for the study performed. It is important that box flow during calibration was identical to mouth flow, and this was ensured by matching the two flows by use of a third-degree polynomial.

The algorithm for calculation of PA was thoroughly tested in the previous paper (10) and found to be adequate for normal subjects. Given that the subjects presented here did not produce results outside the range in which the equipment had been tested, we believe that the measurements are satisfactory for them also.

We have found that the PA continued to increase as flow fell after PEFm was reached (Fig. 3), as we had also found in the normal subjects (10). This indicates that PEFm with most of the added resistors was achieved when Vws limitation occurred somewhere in the airway (2). This is supported by the finding that PEF with added resistances appeared to occur when flow reached the perimeter of the curves for mouth flow, and this was ensured by matching the two flows by use of a third-degree polynomial.

The algorithm for calculation of PA was thoroughly tested in the previous paper (10) and found to be adequate for normal subjects. Given that the subjects presented here did not produce results outside the range in which the equipment had been tested, we believe that the measurements are satisfactory for them also.

We have found that the PA continued to increase as flow fell after PEFm was reached (Fig. 3), as we had also found in the normal subjects (10). This indicates that PEFm with most of the added resistors was achieved when Vws limitation occurred somewhere in the airway (2). This is supported by the finding that PEF with added resistances appeared to occur when flow reached the perimeter of the curves for mouth flow vs. the volume determined by thoracic gas displacement. Measurement of the dwell times at 90, 95, and 97.5% PEFm (DT90, DT95, and DT97, respectively), however, showed that DT90 was not influenced by the mini-Wright PEF meter (Table 2), but larger resistances increased DT90, indicating a blunting of the peak. Blunting was less pronounced for DT95 and DT97 (not shown in Table 2). The increased blunting of the peaks with larger added resistance parallels the decrease in PEF variability, and the two phenomena may be related. The results from the present study fail to confirm the previous finding from a larger study (12) that an added external resistance, such as a mini-Wright PEF meter, reduced PEF variability.

The most marked finding compared with the healthy subjects (10) was the considerably smaller PEFm (3.6 vs. 11.4 l/s) and smaller Ppeak (3.7 vs. 6.7 kPa). One might argue that the smaller PEFm was due to a smaller effort, but this is not true. Studies of isovolume pressure-flow curves in normal subjects and in patients with COPD indicate that PEFm and Ppeak will both decrease with increasing obstruction, and this decrease is independent of the muscle strength (13). The maximal PA in the emphysema subjects (read from Fig. 3) was ~9 kPa, whereas in the healthy subjects it was previously found to be ~15 kPa (10). This discrepancy could be due either to the higher mean age of the patients compared with the healthy subjects (66 vs. 43 yr) or to the chronic pulmonary disease (3).

In the present study, we did not let the patients do complete FVC maneuvers during the repeated blows because that would have been physically very demanding for them. Instead, we asked the patients to relax after 4 to 5 s of maximum effort. The average total

![Table 3. Values that satisfy wave-speed conditions and that are in accordance with model](http://jap.physiology.org/DownloadedFrom/HITP://jap.physiology.org/DownloadedFrom/10.220.33.34.onJuly10,2017)
volume expired, measured at the mouth with and without added resistances, was only 1.41 ± 0.59 liters, and as expected this was much smaller than the pretest values in Table 1 (2.35 ± 0.73 liters).

One of the key questions initiating this investigation was whether a lower Ppeak would diminish gas compression in the lungs of COPD patients and whether this might be counterbalanced by the increased TGV. The increase in PA caused TGV at PEFm to decrease by 0.08 ± 0.10 liters (n = 15). In the previously studied healthy subjects (10), the decrease was 0.24 ± 0.09 liters (n = 10). However, the absolute TLCs were not significantly different between the two groups (6.9 ± 1.6 vs. 7.2 ± 0.9 liters), despite the TLC of the patients being significantly higher than predicted (+1.7 ± 1.5 SR for the patients vs. −0.5 ± 0.6 SR for the healthy subjects). The degree of compression, therefore, is smaller in the patients than in the healthy subjects only because the PA is smaller. It is remarkable, however, that a decrease of TGV by only 80 ml can cause PEF to decrease significantly from 3.6 to 3.2 l/s, i.e., by ~10%.

Our results are in perfect agreement with those of Campbell et al. (1), who in 1957 examined five emphysematous subjects with a mean PEF of 2.3 l/s and a corresponding transpulmonary pressure of 1.8 kPa. They explained this in terms of critical narrowing occurring at lower pressures and more peripherally in patients with emphysema than in normal subjects.

Application of the wave-speed theory of flow limitation (2) to the data from the patients here and the healthy subjects (10) offers a possibility to explore the mechanisms behind the findings. In the APPENDIX, a four-constant parameteric model is described with upstream frictional resistance (Rfr), downstream resistance (Rd), airway compliance (Caw), and airway cross-sectional area at zero transmural pressure (A0) as independent variables, with the assumption of the simplest possible relationship, a straight line, between airway cross-sectional area (A) and transmural pressure (Ptm). Applying a range of Ptm as input, it is possible to calculate corresponding values for the maximum expiratory flow (Vws), Pel, and PA, as described in the APPENDIX regarding the equations shown in Fig. 5. On the other hand, knowing related values of Vws and PA, and assuming values of Pel and Ptm, then Caw, A0, and Rd can be calculated for a range of values of Rfr. Table 3 shows values that satisfy the wave-
speed conditions and are in accordance with the model. For the healthy subjects not blowing through a peak flowmeter, the values for Caw, A₀, and Rd calculated from the model are in close agreement with values measured in healthy young subjects (9). Blowing through the PEF meter is assumed to reduce Pel from 2.0 to 1.7 kPa because of gas compression in the lungs but not to change Ptm at the choke point (CP). For the patients with no PEF meter, it is assumed that Pel is 0.3 kPa (4), that Ptm is −1 kPa, and that blowing through the PEF meter reduced Pel to 0.2 kPa. It is likely that the patient airways are narrower than those of healthy subjects, and because of a more peripheral CP in the patients we assume that Rfr is the same as in the healthy subjects. In that case, Caw becomes considerably larger, A₀ becomes slightly smaller, and Rd becomes larger than in healthy subjects, consistent with a more peripheral location of the CP. For each choice of Rfr, Pel, and Ptm, there is one solution for Caw, A₀, and Rd. Smaller chosen values of Rfr imply smaller values of Caw and A₀ in this model. The Rfr value with the PEF meter was chosen so that both Caw and A₀ at CP were minimally changed, and in that case only a minimal increase in Rfr was seen.

The model correctly predicts an increase in Rd, and this increase is not statistically different from the increase in total resistance actually caused by the PEF meter. It should be noted that Rfr is not identical to the upstream resistance in the terminology of Fessler and Permutt (4), because the latter includes resistance due to convective acceleration of the air in the narrowing airway. Furthermore, their critical transmural pressure (Ptm) is an extrapolated value defining the Ptm when the airway is just closed, and this is not the Ptm at CP. In our model, the relationship between Ptm and cross-sectional area (the “tube law”) is described by a straight line and is only valid for the situation at PEF. Therefore, extrapolation to other flow values should be done with caution. Figure 4 shows the individual relationships between V˙ ws (i.e., V˙max) and PA for the patients and the previously described healthy subjects (10) with and without PEF meter added as resistance. The means for the two groups and conditions are plotted separately. The curves describing the relationships between V˙ ws and PA according to the model in the APPENDIX were calculated from Table 3 and seem to fit the experimental data fairly well, especially for the patients, but similar good fits could be obtained with other combinations of values.

From the wave-speed theory of flow limitation (2) used in our model, it can be predicted that the V˙ max will decrease with 1) lower elastic recoil of the lung, 2) smaller cross-sectional area at the CP, 3) increased Caw at this point, and 4) increased pressure loss upstream of this point.

We conclude that, from among these factors, the lower PEF recorded with the added resistance of a mini-Wright PEF meter found in both patients and healthy subjects is most likely caused by decreased Pel, which is consequent from compression of the TGV and leads to PEF being achieved at a lower lung volume. Although a concomitant increase in peripheral pressure loss due to volume dependence of the airway size is a further possible explanation, this could not be reconciled in the present model. The best explanation seems to be that the decreased Pel leads to a smaller V˙ ws and accordingly a smaller PEF. Therefore, PEF meters should have as low a resistance as possible to avoid this reduction in PEF reading.

APPENDIX

A Four-Parameter Model Describing Flow Limitation During Maximal Forced Expiration

It follows from the wave-speed theory of flow limitation (2) that there are three main factors that determine maximum expiratory flow. These are the Pel, the Pfr upstream of the CP, and the tube law, which relates the distending Ptm to the A at the CP. It is assumed that the tube law is known in terms of a mathematical relationship between Ptm and A, and the simplest relationship is a straight line (see Eq. 1 in Fig. 5). The four parameters in the model are Rfr, Caw at the CP, A₀, and Rd. Studies in humans (9) have shown that these can be derived from measurements of V at the mouth, the esophageal pressure which is taken to represent pleural pressure (Ppl), and bronchial pressures as shown in the upper part of Fig. 5, with Pt being the impactation pressure and Plat the lateral airway pressure.

The straight-line tube law (Fig. 5, bottom; Eq. 1) is defined by the Caw and the A₀. Equation 2 in Fig. 5 defines the relationship between the V˙ ws and the tube law (Eq. 1) where ρ is the gas density. Equations 3 and 4 define the pressure losses due to Rfr and Rd. The pressure drop due to convective acceleration (Pca) is defined in the Bernoulli equation (Eq. 5). Equations 6 and 7 are derived from the pressure diagram in the upper part of Fig. 5, by applying Kirchoff’s law, and Eqs. 8 and 9 follow directly from this diagram.

The parameters in boldface are those that define the model. By solving a number of equations for a series of possible Ptm values, it is possible to calculate relationships among V˙ max and Pel, Ppl, Ptm, Pfr, Pa, the pressure head (J) at the CP, and the pressure loss (Pd) due to Rd. In this way the relationships between V˙ ws (i.e., V˙ max) and Pa shown in Fig. 4 were calculated. The model is always valid for the Ptm at which the parameters are derived, but the model can to a certain extent be applied to explain flow limitation under different circumstances.

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